A NEW BIOMECHANICAL ASSESSMENT OF MILD TRAUMATIC BRAIN INJURY
PART I - METHODOLOGY

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ABSTRACT
Mild traumatic brain injury (MTBI) occurs daily in virtually every human activity. Certainly, traffic accidents account for the vast majority of all such trauma. Concussions also occur in sports and recreational activities. American professional football provides a unique opportunity to study MTBI. That opportunity comes not from the particularly high frequency of occurrence (many sports are higher, e.g. boxing, ice hockey), but from the following facts:

- The game is played on a precisely laid out grid.
- The movement of players is recorded by video cameras from a multitude of directions and angles.
- The subjects are all of comparable stature, age and physical condition and they are all equipped with virtually the same protective headgear.
- Physicians at the sidelines are available immediately to diagnose and treat an injury.

This unique "laboratory" is being analyzed through a program involving four complimentary approaches:

1. Video recordings of MTBI events are analysed to determine the kinematics, i.e. the relative speed, direction and impact sites on player's heads.
2. In addition to clinical observations made by team physicians, concussed and uninjured players are subjected to neuropsychological assessment.
3. Re-enactments of certain collisions are conducted with instrumented anthropometric test devices (ATD's).
4. Responses from the ATD heads are used to drive a mathematical model (FEM) of the human brain that predicts intracranial distortion patterns.

It is intended that the results of the neuropsychological testing, as well as the medical evidence, be correlated to the FEM brain distortion patterns. The overall objective of the program is to develop new biomechanical criteria for MTBI. The new criterion functions will be based upon the measured ATD head translational and rotational accelerations and will be helpful in developing new standards for protective headgear. The present paper outlines the methodology and considers one example of incident reconstruction.

THE DATABASE

The National Football League NFL has sixteen scheduled games during the regular season. There are 30 teams (each with up to 50 players). During the season, all games are videotaped and televised. Each year, approximately 150 players are diagnosed as having sustained a possible mild traumatic brain injury MTBI during a game. In each case, the team physician conducts an examination and often a follow up exam is conducted. Clinical and somatic symptoms are documented and the data is fed into a large database. This database is maintained in order to consider the epidemiology of MTBI (Powell, 1998). In addition to the clinical studies, many teams participate in an independent neuropsychological test program: (Lovell, 1998). This program is aimed at developing a quantitative measure of when a player returns to his pre-injury neurological status. Such a tool is extremely valuable in deciding when a player might be ready to return to play. Finally, an advanced neuropsychological test program, that is endeavoring to create a "Concussion Severity Index", is underway with five selected teams. This work is described in somewhat greater detail below.

When a player is first diagnosed with a possible MTBI, the time during the game when the incident likely occurred is documented by the team training staff. Videotape recordings of the game during that time interval are reviewed at NFL headquarters. In those cases where it is reasonably clear in which specific play the incident apparently occurred[1], all videotape recordings of the game during that interval are culled from the broadcast tapes for later detailed analysis.

KINEMATIC RECONSTRUCTION

The game of American football is played upon a field made of natural or artificial turf. The field employs a grid that clearly identifies location and distances on the playing surface. A football field defined by the NFL has the dimensions and markings shown in Figure 1.

[1] It is not always the case that the suspected MTBI is clearly associated with a specific event on the field of play.

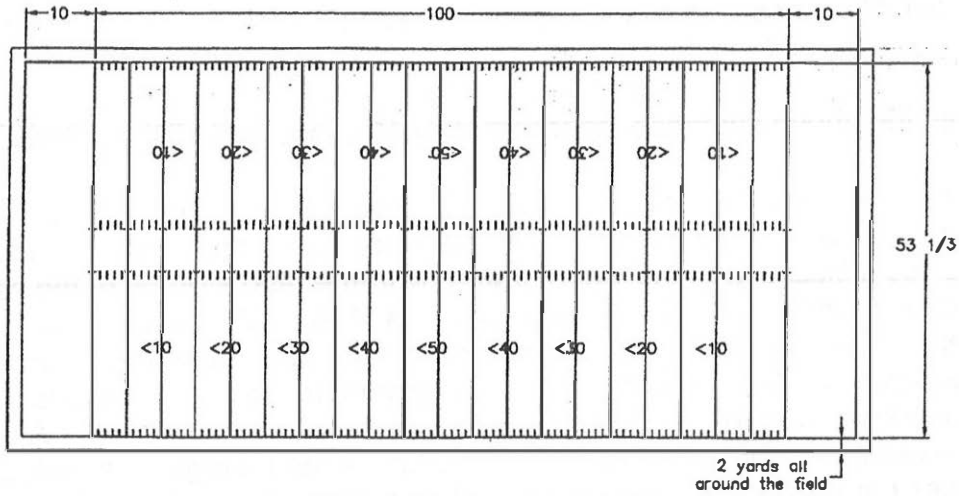

Figure 1: Dimensions of a NFL Football Field

Several cameras are placed around the playing field, some are fixed in space, and others are hand held and rove during the game. The fixed cameras are typically in the stands and provide an overhead view. The hand-held cameras are typically at field level and often provide close up views. All cameras can change effective focal lengths, i.e. they "zoom" and they also usually "pan", resulting in a view that is not stationary with respect to any field reference point.

Nearly 100 plays during the 1995-1998 seasons, where a player apparently sustained a concussion, were carefully reviewed. In most cases, several camera views were available. In principal, it is possible to extract reasonably precise player movements by analyzing videotapes from multiple camera views of the same incident. The difficulties stem from not knowing the camera locations or the degree of magnification each camera provides. Having cameras with different framing rates compounds the problem. Inadequate resolution and speed blur adds further to the difficulties. For these reasons, not all videotaped plays that are apparently associated with a concussion are suitable for detailed analysis.

In order to reconstruct any of these events, two or more camera views are required, each including the following important criteria:

- Head-to-head contact without additional involvement from shoulders, arms, etc.
- The head/helmet(s) should be clearly visible at contact as well as having approximately five video frames prior to contact.
- The relative angle between the cameras should be between 30 to 150 degrees for sufficient vector resolution.
- Field references must be visible (yard lines, side line, "hash marks", etc.) that allow alignment of a reference co-ordinate system near the point of contact.

- The helmet(s) in the view of the camera should be sufficiently large to establish a scaling factor, as well as to distinguish the orientation and contact points of the helmets.

From the above database, approximately 15 cases met all the criteria[2].

In order to determine the pre-impact kinematics and the nature of the head interaction in any of these cases, the videotapes are first digitized. The process involves digitization of the video, frame by frame, and storage of the data on a computer hard drive. The system used is a MIRO video digitizing board operating at 7.76 MIPS which produces 30 full frame captures per second with a resolution of 720 x 640 and a colour depth of 16 bits. This provides an MPEG compression ratio of 2.5 with no lost frames. An ultra wide SCSI 2 A/V hard drive easily accepts this data rate. Standard video footage is recorded at 30 frames per second thus the above digitizing does not result in video frame loss. However, each video frame is a combination of 2 fields that are recorded at 60 fields per second. To digitize each individual field, the real-time video footage was played back at half speed (30 fields per second instead of 60 fields per second) such that each field would be digitized as if it were a frame. In doing this, the extent of speed blur was reduced and the time base resolution was increased by a factor of 2.

The following steps are then undertaken:

- Determination of camera viewing angles.
- Image scaling.
- Tracking of images relative to fixed or moving reference image.
- Recombining relative velocity vector components.
- Determining contact points on players' helmets.

DETERMINING CAMERA ANGLES

A method for approximating the camera positions relative to the football field was developed using digitized video images of a documented incident that was associated with possible mild traumatic brain injury. Reference marks on the field and a computer-drafting package (AutoCAD®) were used to determine the camera angles relative to the field position of the incident. The steps involved in establishing camera angles are detailed below.

A set of three lines, representative of three orthogonal lines or axes on the football field, were superimposed on a digitized image at or near the site of impact. These lines were drawn such that one line was vertical, another was parallel to the yard lines on the field and the third was parallel to the sidelines. Reference marks on the field, such as "hash marks", yard lines, sidelines and yard numbers aided in establishing these axes. To minimize the effects of

[2] An additional important criterion is that the clinical and neuropsychological data for the players involved be available for analysis. Though many players are tested, the data is confidential and they are under no obligation to provide this information to the research effort. Those players who do agree to make their records available will still have confidentiality preserved, as they will not be identified in this or any other publication.

perspective, the three axes were drawn as close as possible to the region of the digitized image where the incident occurred. An example of a set of three orthogonal lines superimposed on a video image of an incident is illustrated in Figure 2. The two dimensional planar view of the three orthogonal axes was recreated in AutoCAD's paper space. While in paper space, a window was opened looking in on an accurately scaled three dimensional representation of a NFL football field.

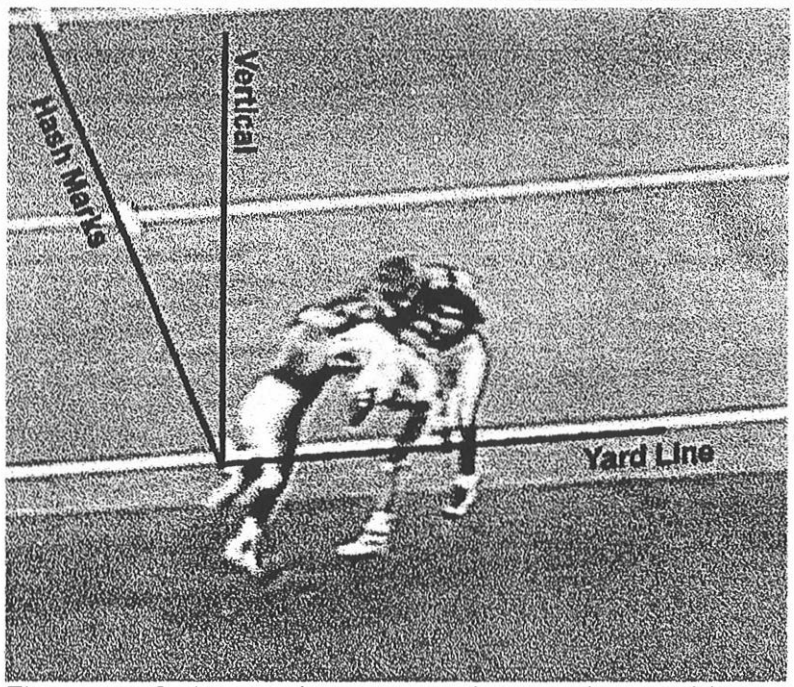

Figure 2: Orthogonal axes superimposed on a video image of incident

The three dimensional viewing angle of the football field was adjusted such that the reference marks on the field were aligned parallel to the corresponding axes drawn in the previous step and shown in Figure 3.

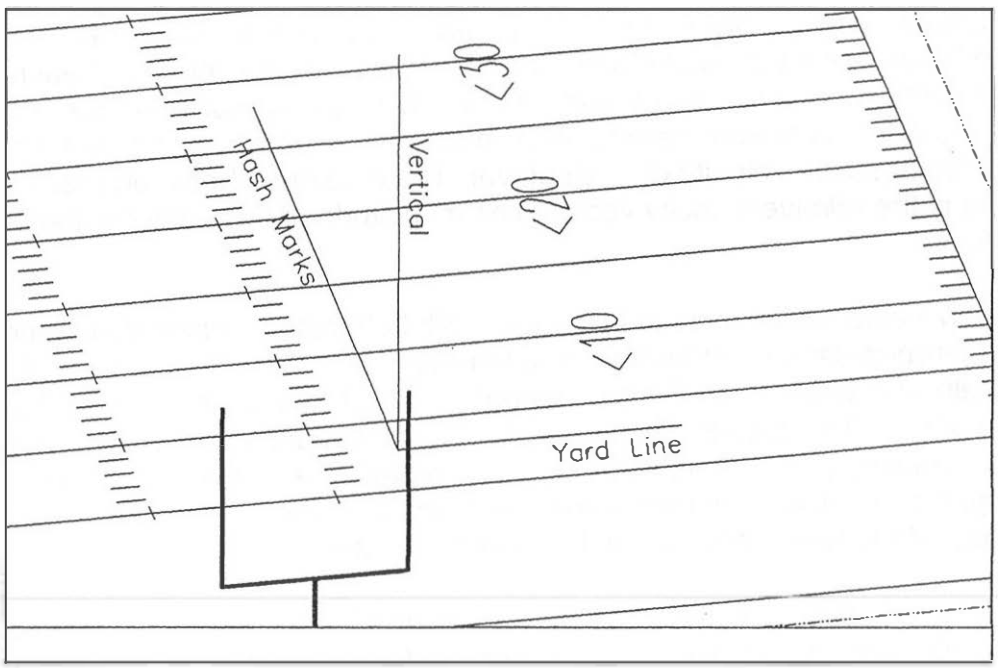

Figure 3: Two-dimensional representation of orthogonal axes superimposed on an AutoCAD® football field

Once alignment between the axes and the reference marks on the field was achieved, the software provides the direction from which the football field has been observed[3]. The angles for three cameras in the case example are shown in Figure 4. Two of these viewing angles were used in determining the impact velocity.

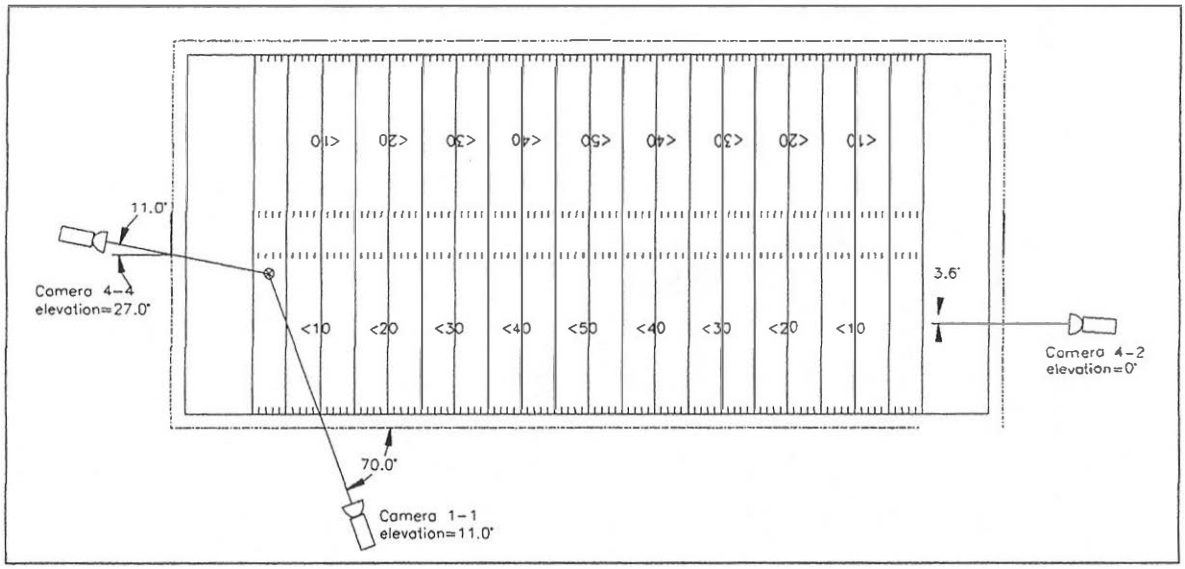

Figure 4: Camera angles for example case

DETERMINING RELATIVE VELOCITY

There are two approaches in determining the speed with which two players move relative to each other. Wide-angle views of the playing surface upon which both players are moving provide the most direct method. The position of each player's head as a function of time (established from framing rates and the appropriate scale factor) is measured relative to some field reference point. This method provides a (film plane) component of the absolute velocity of each player. If the two players are assumed to be moving in the plane of the playing field, and once a camera-viewing angle has been established, this method yields the absolute velocity of each player. The difference between these two vectors is the relative velocity vector. The magnitude of this vector is the relative speed.

Close up views, which include images of both helmets but which do not include field reference data, must be treated differently. In these cases, a central point on a helmet is tracked relative to a co-ordinate system whose origin is at the centre of the other helmet. This process yields a component (film plane) of the relative velocity vector. This same process, repeated with another view (with appropriate consideration for scaling), provides the other required component. The sum of the two yields the relative velocity vector.

The relative impact velocity of two players involved in the example case was calculated using the digitized video footage of the incident and from the

[3] The direction provides the camera viewing angles but not the distance of the camera from the field.

knowledge that each subsequent video field, that comprises a video frame, is a 60th of a second from the previous field[4]. The relative impact velocity of two players involved in a collision was calculated using the digitized video footage of the incident and from the knowledge that each subsequent video field that comprises a video frame, is a 60th of a second from the previous. Image analysis software (Inspector by Matrox®), was used to determine the relative distance between the two colliding helmets. The distance was measured by converting the number of pixels separating the helmets to units of meters using the size of the helmet for scaling. The change in the relative position of one helmet to the other over time results in the relative velocity between the two helmets, as measured in the frame of view of that particular camera. The same procedure was repeated for a second camera position, resulting in a relative velocity measurement from the additional view.

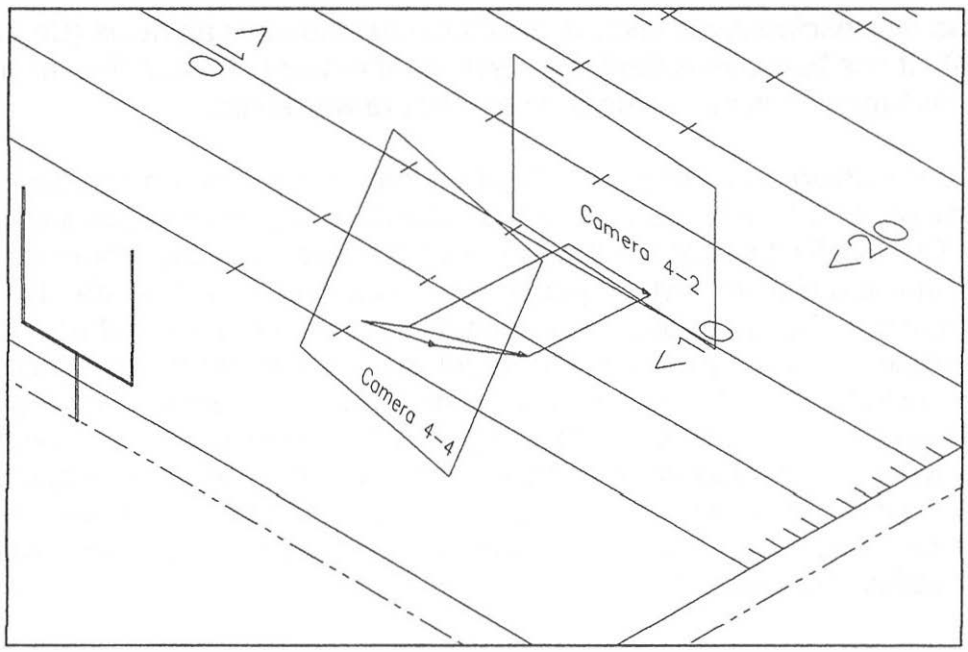

Figure 5: Graphical reconstruction of the relative impact velocity

The absolute relative velocity was graphically determined by combining the relative velocities that were measured in the two different camera views. This was accomplished using AutoCAD®. Both camera planes were established relative to the football field using the camera angles determined previously. In each of the camera planes the relative velocity (magnitude and direction), established using the image analysis software, was drawn. These vectors were then projected back towards the plane of the football field, perpendicular to the camera planes. The velocity vectors were positioned in their respective camera plane, such that, when projected onto the plane of the football field, their starting points coincided. The end point of the relative velocity vector was established by the intersection of the projected end points from the camera

[4] Post-processing was required to eliminate additional duplicated frames that are inserted in video footage to produce slow motion replay. Half speed replays could be digitized as indicated above, however, slower than half speed replay or variable speed replay required that all the duplicate fields be removed manually.

views. The length of the resulting vector becomes the magnitude of the relative velocity. The graphical reconstruction of the relative velocity for the case example is illustrated in Figure 5. The recombination of camera plane relative velocities for the case example resulted in an absolute relative impact speed of 10.4 m/s.

NEUROPSYCHOLOGICAL EVALUATION

Concussion has been defined as a traumatically induced alteration in mental status with or without loss of consciousness (Kay et al, 1997). Attempts to quantify the severity of a mild traumatic brain injury have usually been based upon clinical observations. According to one classification (Colorado Concussion Scale) a grade 1 concussion is defined as confusion only with no amnesia. A grade 2 concussion includes amnesia, and a grade 3 concussion involves any loss of consciousness. Other grading scales include the duration of loss of consciousness and the extent of posttraumatic amnesia (Cantu, R.C., 1996). It has become evident, for a variety of reasons, that additional, more objective measures of concussion severity are warranted.

Neuropsychological testing is a field of endeavour that attempts to quantify the degree of neurological dysfunction that an individual possesses at some point in time. Classically, this has involved the application of tests that compares the individual to a similar "normal" population. These methods typically are used to measure the effects of possible congenital defects, disease or other neurological disturbances. They have not been developed to be especially sensitive to concussion per se. In fact, only recently have attempts been made to identify a specific battery of neuropsychological tests that do appear to be sensitive to MTBI (McCrea et al, 1998). At this point in time, there appears to be no general consensus as to what are the "best" tests. What is known is that the neurological behaviour of one who has sustained trauma to the head can be observed and measured.

In the current program, a new computer-based battery of neuropsychological tests is being developed. The battery initially comprised the following cognitive tests:

- Symbol Digit Modalities (SDM)
- Colour Trails Tests (CTT)
- Simple Reaction Time (SRT)
- Choice Reaction Time (CRT)
- Visual Tracking (VT)

The neuropsychological battery was implemented in custom-designed software on a personal computer. The computer includes a data acquisition card with on-board timers as well as a data tablet for pen input. The hardware/software system design was optimized to enable subjects to perform the SDM, CTT, SRT and CRT tests using standard protocols for comparison with existing normative data. However, in addition to the measures normally available from these tests, the current system also records additional details of the subject's response in a manner that is transparent to the test subject. Hence, more detailed response information is provided without altering the tests.

Players are provided verbal instructions and are then allowed to interact with the programs using the pen of the digitizing tablet and/or the keyboard. Responses and interval times are recorded using operating system calls as well as a digital counter from a data acquisition board. Results are stored to disk after each test, and are backed up after a testing session is completed.

The tests were conducted and data collected from players on five different NFL teams. With the exception of visual tracking, the baseline (i.e. pre-injury) data were normalized for 208 players. The selection of cognitive measures was based upon the observation that the underlying basis for many of the frequently cited cognitive symptoms associated with mild diffuse brain injury (attention, memory and concentration difficulties) is actually an impairment in information processing. The cognitive tests used in this battery are time dependent tasks that are sensitive and specific for assessing processing speed capacity.

In addition to the team physicians' clinical assessment and the neuropsychological tests, subjects who were concussed are assessed regarding their awareness of time and date, and each also completed a Functional Status Checklist (FSC-a self-reported rating of symptoms commonly associated with concussion). The details of all these test methodologies are reported elsewhere (Thibault et al, 1999).

In the case example presented here, the concussed player completed the battery of neuropsychological tests three times: pre-injury, after the concussion (1 day post injury), and during a follow-up examination (5 days post injury). His results, which are reported in greater detail elsewhere (Thibault et al, 1999) suggest that the reaction time tests SRT and CRT, as well as the individual's own assessment (FSC) are more sensitive and specific in identifying cognitive deficits than any of the other tests. In this case, 1-day post injury, the individual's reaction times were markedly lower than the baseline values. Five days post injury the reaction times had recovered somewhat but they had not returned fully to their pre-injury levels. These observations by themselves do not provide a MTBI "severity scale" as such, but when combined with clinical observations and other neuropsychological data, they certainly point the way toward such an objective measure.

INCIDENT RE-ENACTMENT

The precise form of the laboratory-based reconstruction depends on the circumstances of the particular incident. In general, a Hybrid III ATD head represents the head of a player. The ATD "wears" a football helmet of the same make and model that the player wore during the incident. In the case of two colliding players, two helmeted headforms are used. Either or both may be instrumented with nine linear accelerometers. The ATD heads are attached to Hybrid III necks. Either or both may be instrumented with typical 6 axes load cells.

From the kinematic video analysis, the velocity of one player relative to the other is determined. In most cases both players are moving. However, in the laboratory, it is not convenient to move two headforms towards each other.

Therefore, one player is represented as a stationary headform, and the other is moved at the full relative velocity.

In order to reproduce the impact, one head-neck assembly is mounted to a rigid base, and the other to a vertical-guide free-fall system[5]. The assemblies are aligned to correspond as closely as possible to the observed pre-impact configuration. The orientation of the colliding headforms is judged by careful review of the videotapes, looking at all available camera angles. One of the headform assemblies is then raised above the other to a height that upon release in free-fall yields (by gravity alone), the intended impact speed.

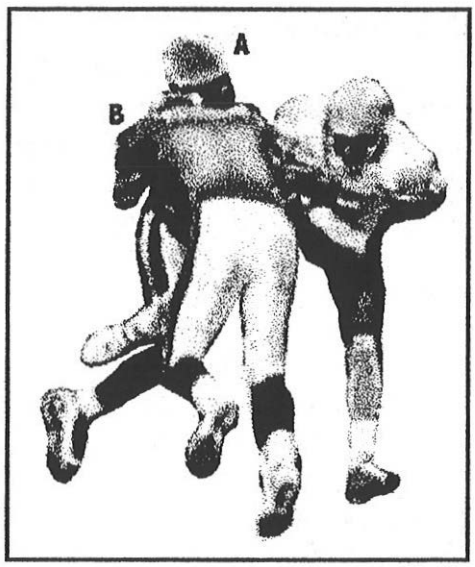

Figure 6: Sample case, view in line with impact

For the example case here, the MTBI football game incident was as follows. Two players of the same team were attempting to tackle a pass receiver, but from opposite directions. The pass receiver avoided the tackle, and the two tackling players ran headfirst into each other. Their heads collided, with player A being hit on the right side of his faceguard by the front crown of player B. Player A sustained an apparent MTBI, while Player B did not.

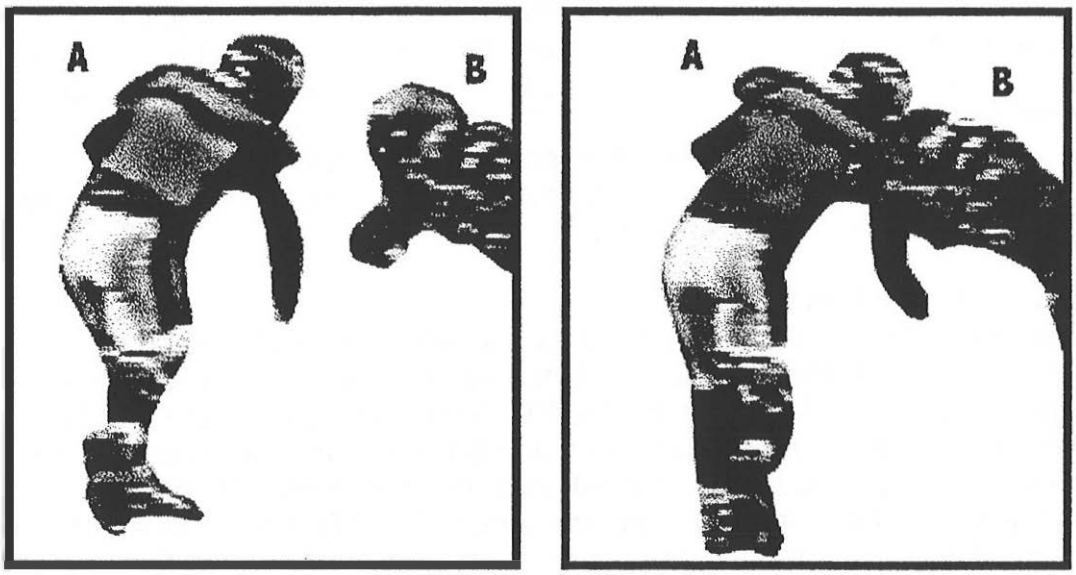

Figure 7: Sample case, side view of impact

[5] The impact, which in reality often occurs in essentially the horizontal plane, is reproduced in the vertical plane.

Figure 6 shows the impact orientation of players A and B, viewed from the rear of player B. The player with the light-colored jersey was the pass receiver. In Figure 7, the same impact is shown from the right of player A (left of player B), but with the pass receiver omitted for clarity. The background of these figures was also removed for clarity. Note that both figures were captured from video footage.

Video footage showed that when the head of player A was hit, his chin was deflected inwards, and then deflected towards his left shoulder. Torso motion was almost unaffected by the initial impact until the head and torso of player B made contact. The motion of player A's head was considered to be uncoupled from his torso for the duration of the helmet-to-helmet impact. This was represented in the laboratory by fixing the base of player A's neck to a rigid base and impacting it with a representation of player B's head[6].

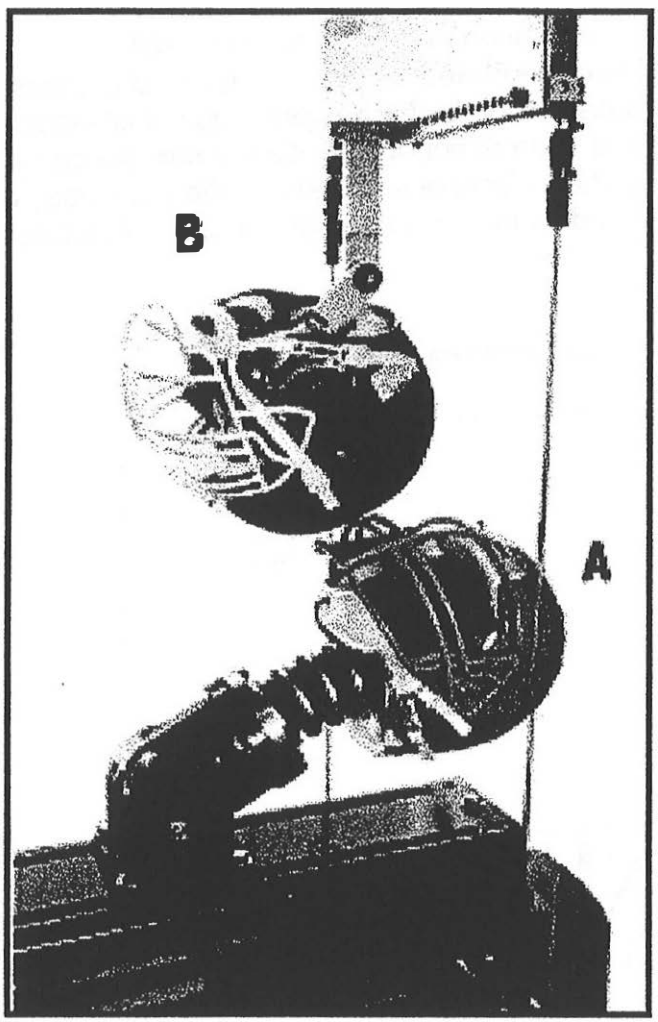

Figure 8: Laboratory re-enactment set-up.

The test set-up is illustrated in Figure 8. The headform, which represents player B, is suspended from an adjustable arm is suspended from a wire-guided carrier. Upon impact, both headforms A and B were able to deflect in a manner similar to that seen on the video. The reader is reminded that the impact here is in the vertical direction, rather than horizontal as on the football field. Also, the illustration in Figure 8 is from the left of player A, unlike Figure 7, which is from the right of player A.

[6] This initial re-enactment attempted to preserve the impact orientation and speeds as witnessed in the game video. However, in future re-enactments, it is anticipated that more attention shall be needed in the area of momentum conservation. While it is important that headform mass be correct, it is also important that the effective torso mass be considered. Future re-enactments shall endeavor to better account for representative torso mass in the testing.

Headform B was raised to a height that yielded an impact speed of 10.4 m/s, which was the speed calculated from the video analysis of this incident. Video records of this re-enactment illustrated that the post impact kinematics of both heads were similar to that seen in the game video.

Only headform A was instrumented in this re-enactment. Peak resultant acceleration of the centre of gravity was 52 G. Peak resultant rotational acceleration was 5600 rad/sec^2. The data traces for linear and rotational headform response are shown in Figure 9 and Figure 10.

Historically, headform responses of the form above have been used to delineate the likelihood or severity of a head injury. That is, some function of the magnitude of the resultant linear acceleration and/or of the rotational acceleration has been correlated to closed head injury severity. To date, these functions have not been entirely satisfactory. In the present study, it is expected that more plausible functions, based upon direct observation of mild traumatic brain injury will evolve. To develop these functions, the kinematic responses of the "injured" headform are being used as input into an advanced mathematical model of the human brain.

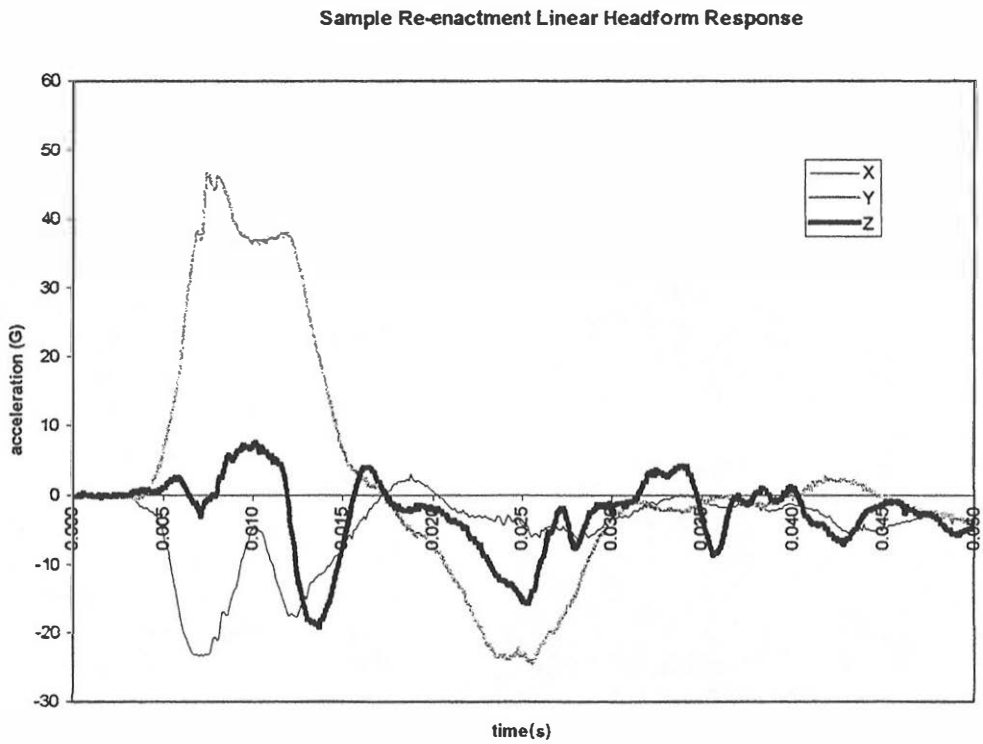

Figure 9: Re-enactment linear headform response

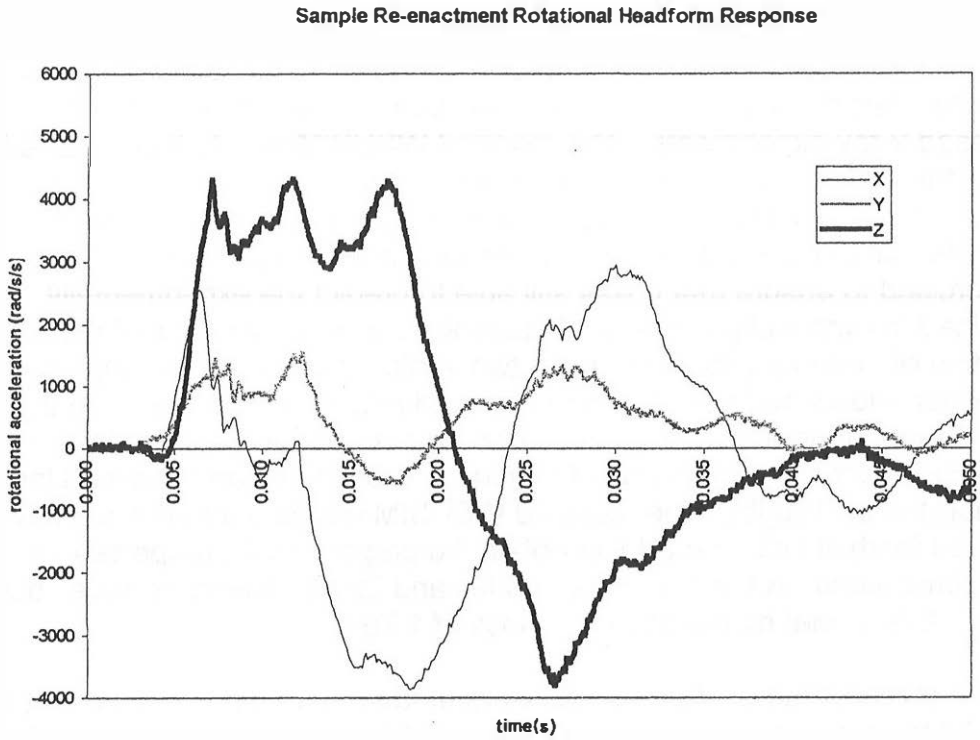

Figure 10: Re-enactment rotational headform response

FINITE ELEMENT MODELING OF THE HUMAN BRAIN

Under circumstances of moderate impact, a helmet renders the kinematic response of a human head essentially that of a rigid body, i.e. there is negligible skull deformation. Injuries that can occur do so solely as a direct result of inertial loading brought about by movement of the head as a whole, not through skull bending.[7] As such, the modelling exercise becomes one of simply driving the skull of the FEM to move exactly as the ATD headform moved during the simulation impact. To model the responses of the players' brains during the impact, the six acceleration components of each of the ATD heads are used as input for the Wayne State University Brain Injury Model (WSUBIM).

MODEL DESCRIPTION: The finite element model used for this incident reconstruction was an updated WSUBIM. The model comprises most of the essential components of the head including scalp, skull, dura, falx cerebri, tentorium, pia, cerebrospinal fluid (CSF), venous sinuses, ventricles, cerebrum (gray matter and white matter), cerebellum, brain stem and bridging veins. The model geometry represented the head of a 50th percentile adult male. The model is meshed with 28,754 nodes and 37,040 elements. The total mass of the model was 4.22 kg with 1.45kg contributed by the brain.

The original WSUBIM was partially validated against the intracranial pressure data from one of the cadaver frontal impact tests reported by Nahum et al, (1977). Recently, it was refined to include a 3-layered skull and was used to

[7] This may not be true in the case of very high-energy impacts but certainly would be so in those that occur on the football field.

simulate all experiments reported in the study by Nahum et al, (1977). The peak intracranial pressure as a function of total impact energy predicted by this modified model matched well with those obtained experimentally (Al-Bsharat et al, 1999a). Additionally, the model was validated using data obtained from the high-speed x-ray experiments. The modified WSUBIM was further modified to improve the ability of the brain to slide relatively freely with respect to the skull. A sliding interface between the skull and the CSF layer was introduced to replace the direct connection between the skull and the CSF. The model was first exercised to ensure that it was still able to predict the experimentally determined intracranial pressures. Experiments were designed to determine the motion of brain targets arranged in two vertical rows, using the high-speed x-ray system described above. The measured relative motion between the skull and the brain during a blunt impact was compared to that predicted by the model. Excellent correlation was obtained for relative motion of several targets (Al-Bsharat et al, 1999b). The modified WSUBIM was integrated into a newly developed football helmet model to obtain the proper inertial properties for impact simulations. A total of 24,140 nodes and 22,532 elements were used to represent the helmet model having a mass of 1.75 kg.

The distinctions of material properties were made among each component of the model to represent the anatomic features of the human head (Table 1). The published experimental data for mechanical characteristics of brain tissue vary over a broad range. The actual brain properties remain uncertain. However, it is generally accepted that the brain tissue is a highly damped visco-elastic material. In this study, the visco-elastic law of constitutive equation was chosen to model the brain tissue. The visco-elastic material behavior was characterized as visco-elastic in shear with the deviatoric stress rate dependent on shear relaxation modulus, while the compressive behavior of the brain was considered as elastic. The shear characteristics of visco-elastic behavior of the brain was expressed by:

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t}$$

where, G_0 is the short term shear modulus, G_∞ is the long term shear modulus, β is the decay constant and t is the duration. The shear modulus values used in this study were derived from Shuck and Advani, (1972).

The distinctions of shear modulus of brain tissue were made between gray and white matter. It is thought that in the region where a majority of tissues are nerve fibers, it would be tougher than in those regions that comprised mostly of nerve cell bodies. The experimental study of shear modulus of the porcine brain tissue also revealed regional differences in mechanical properties of central nervous system (Arbogast and Margulies ,1997). The reported shear moduli of porcine brain stems were 20% greater than those of the cortex at strains of 2.5%. In the current model, shear modulus of white matter was assumed to be 20% higher than that for gray matter, while the same decay factor and bulk modulus of 2.19 GPa were used for both gray and white matters.

The material properties of cranial bone used in this study are based on the experimental data from McElhaney et al, (1971). An equivalent Young's modulus of 8 GPa and a strain of 0.8% at yield were assumed for the skull bone to follow the elastic-plastic constitutive equations. All other tissues of the head were represented as homogeneous and isotropic elastic materials.

IMPACT SIMULATIONS: The acceleration components obtained during the impact re-enactment and depicted in Figure 9 and Figure 10, were used to prescribe the kinematics of the WSUBIM. The linear and rotational accelerations components measured about the CG of the ATD head were applied to a fixed point on the model, approximating the head CG location.

The pressure response of the brain is illustrated in Figure 11 with the peak Coup pressure coinciding with the peak linear resultant acceleration. The Contrecoup pressure lags by 3 ms, likely due to the combined loading effects of the translational and rotational kinematics.

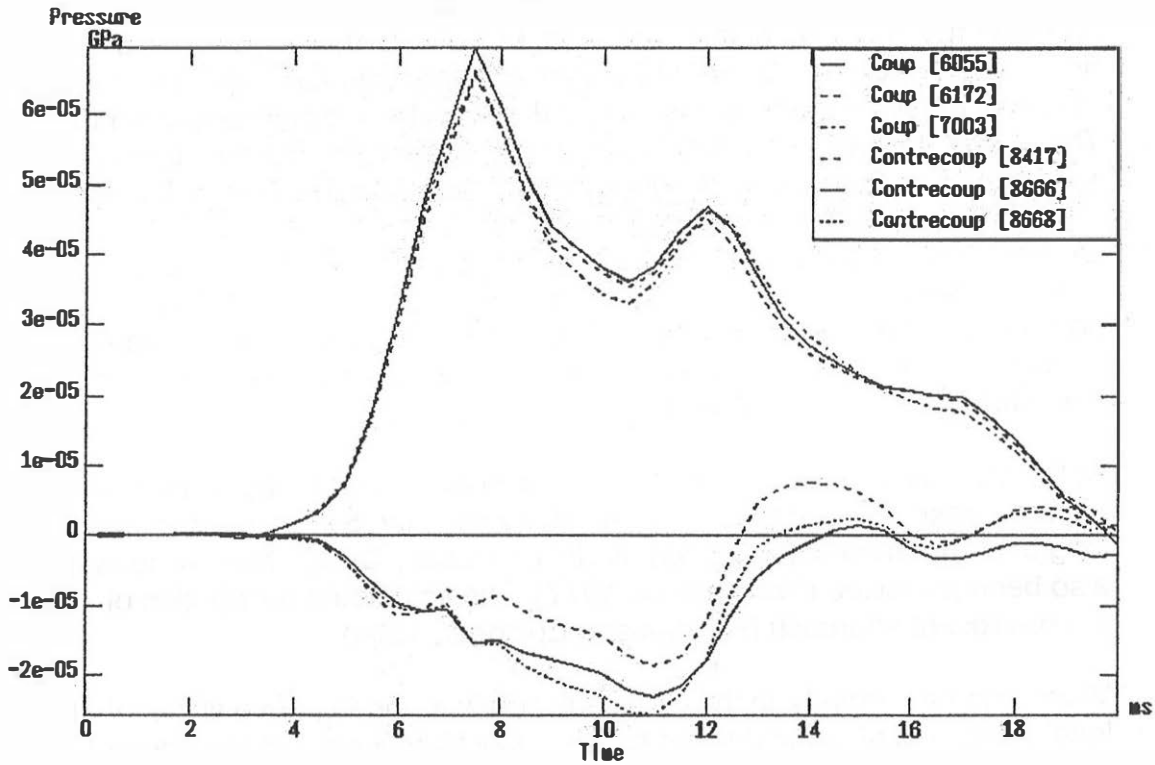

Figure 11: Coup and Contrecoup brain model pressures

The shear stress response of the model is depicted in Figure 11 at time frames corresponding to the peak translational accelerations (8 ms) and peak rotational accelerations (20 ms). Areas of high shear stresses are noted at the periphery and centre of the brain.

The pressures, stresses and strain patterns of the brain are to be used for characterization of the brain's response under various loading conditions associated with concussive and non-concussive events. Direct comparison of

the responses, or derivatives thereof, with the clinical and neuropsychological assessments will aid in the understanding of brain response and development of injury criteria.

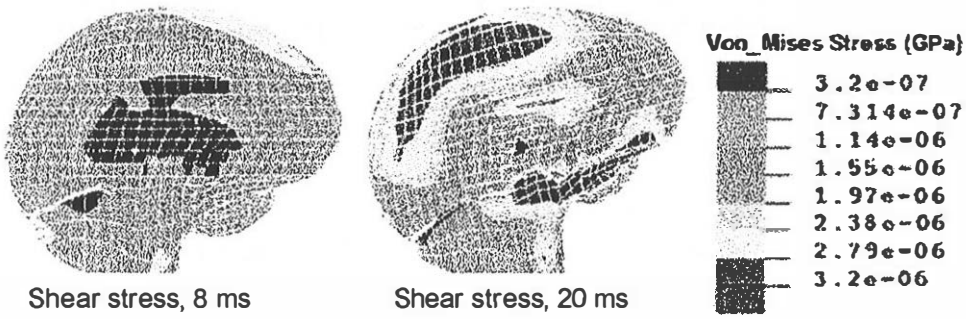

Figure 12: Brain model response to combined rotational and translational kinematics

MILD TRAUMATIC BRAIN INJURY CRITERIA DEVELOPMENT

In order to develop a more appropriate MTBI criterion function, the task is to correlate the distortion of the FEM brain to the rigid skull kinematics of the ATD head. The basic premise is that the inertial movement (i.e. the acceleration) of the head resulting from impact causes the brain to be deformed within the skull. The challenge is to find a direct correlation between the brain distortion and the skull movement and relate these to the pathophysiology of brain damage.

On a macroscopic level, a number of acceleration-based head injury criterion functions have been developed over the years. Newman, (1998) has provided a detailed summary of most of these. Logically, most of these are dependent upon linear and/or rotational acceleration and in some cases, a measure of the time duration of the acceleration.

At the local tissue level, there are several models of brain distortion that have been proposed as correlates to brain injury severity. Simple shear strain has long been regarded as one such model (Hodgson, 1979). Normal stress has also been proposed (Nahum et al, 1977). The volumetric distribution of strain is a more recent approach (Bandak and Eppinger, 1994).

Viano and Lau, (1988) in their "viscous criterion" model, have shown that the local product of strain and strain rate is a key biomechanical parameter for diffuse brain injury severity. In the context of any viscoelastic organ undergoing dynamic deformation, the key biomechanical parameter for injury is the product of the velocity of deformation V and the nondimensional compression C, i.e. the "viscous criterion VC". This term, they point out, certainly for viscoelastic systems such as the brain, is a measure of energy dissipation by dynamic deformation. Thus the total energy dissipated in the brain, as a consequence of its inertial response to an impact, is a direct measure of the probability and/or severity of diffuse brain injury. This approach has recently been propounded again by Viano and Lovsund (1999).

The hypotheses in the present work are:

- The most sensitive indicator of MTBI severity will, at the tissue level, be the product of strain and strain rate. Macroscopically, the viscous criterion VC will determine the brain injury risk Globally, the parameter most sensitive to diffuse injury throughout the brain, will be the magnitude of the total energy absorbed/dissipated by the brain as a consequence of the (rigid body) movement of the skull initiated by the head impact.
- Higher severity of MTBI will correlate to a higher level of energy dissipated.
- The energy absorbed by the brain will be correlated to the three dimensional motion of the head.

In particular, it is proposed that the kinematic criterion function, to which the brain internal energy increase will be correlated, will be a variation on the weighted head acceleration model proposed by Newman, (1986) the GAMBIT[8]. When originally introduced, it was a time dependent non-linear weighted sum of the linear and rotational acceleration of the head. The model had a certain empirical appeal and was substantiated by the then existing biomechanical data. Though means to include the effects of time were included, GAMBIT did not include time or time duration as discreet independent variables. A more simplistic approach will now be considered. It is proposed to modify the original GAMBIT to incorporate average translational and rotational accelerations and time durations as separate independent variables[9].

The first stage of the correlation exercise will be to run the FEM brain for a series of hypothetical time dependent translational and rotational acceleration combinations. The average accelerations and time durations will be fixed and noted. Several such combinations with different spatial orientations, averages and durations will be considered. In each case, the magnitude of the increase in brain internal energy which exceeds some limiting value, will be computed. These values will then be correlated to the corresponding values of the revised GAMBIT formulation.

As more cases of MTBI are reconstructed, and as more suitable measures of MTBI severity become available, the observed brain injury severity will be correlated to the numerical values of both the microscopic(local) and macroscopic(global) criterion functions.

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[8] Generalized Acceleration Model of Brain Injury Tolerance

[9] This is in keeping with the classical Wayne State Tolerance Curve approach upon which so many helmet standards are in some way, based.

Table 1: Material properties used in the FE head model

Substructure	ρ [kg/mm^3]	E [GPa]	ν [-]	K [GPa]	G [GPa]	G_0 [GPa]	G_∞ [GPa]	β [s^{-1}]
Skull	1.21E-06	8.0	0.22	4.76	3.279			
White Matter	1.04E-06			2.19		4.1E-05	7.6E-06	700
Gray Matter	1.04E-06			2.19		3.4E-05	6.3E-06	700
Brain Stem	1.04E-06			2.19		4.1E-05	7.6E-06	700
CSF	1.04E-06			2.19	5.0E-06			
Ventricle	1.04E-06			2.19	5.0E-07			
Dura, Falx & Tentorium	1.133E-06	0.0315	0.45					
Pia	1.133E-06	0.0115	0.45					
Facial Bone	2.10E-06	5.54	0.22					
Skin	1.2E-06	0.0167	0.42					
Bridging Veins	1.133E-06	0.0002 N/mm						

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